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METHOD AND APPARATUS FOR ADJUSTABLE AVD PROGRAMMING USING A TABLE

Cross Reference to Related Applications

This application is related to commonly assigned, co-pending U.S. Patent Application No.10/314,910, entitled "METHOD AND APPARATUS FOR OPTIMIZING VENTRICULAR SYNCHRONY DURING DDD RESYNCHRONIZATION THERAPY USING ADJUSTABLE ATRIO-VENTRICULAR DELAYS," filed on December 9, 2002, and U.S. Patent Application No. 10/314,899, entitled "METHOD AND APPARATUS FOR OPTIMIZING STROKE VOLUME DURING DDD RESYNCHRONIZATION THERAPY USING ADJUSTABLE ATRIO-VENTRICULAR DELAYS," filed on December 9, 2002, which are continuations-in-part of U.S. Patent Application No. 10/243,811, filed on September 13, 2002, which is a continuation of U.S. Patent Application No. 10/008,830, filed on December 7, 2001, which is a continuation of U.S. Patent Application No. 09/661,608, filed on September 14, 2000, now issued as U.S. Patent No. 6,351,673, which is a continuation of U.S. Patent Application No. 09/492,911, filed on January 20, 2000, now issued as U.S. Patent No. 6,360,127, which is a continuation of U.S. Patent Application No. 09/075,278, filed May 8, 1998, now issued as U.S. Patent No. 6,144,880, the specifications of which are incorporated herein by reference.

Field of the Invention

The present invention relates generally to a method and apparatus for cardiac pacing and, in particular, to a pacing system providing adjustable atrio-ventricular time delays to improve various heart performance parameters.

Background of the Invention

The heart is the center of the circulatory system. It is an organ which performs two major pumping functions and may be divided into right and left heart "pumps." The left heart pump draws oxygenated blood from the lungs and pumps it to the organs of the body. The right heart pump draws blood from the body organs and pumps it into the lungs. For a human heart, the right heart pump is on a patient's right side and the left heart pump is on the patient's left side. Figures in this document, such as FIG. 1, show a "top" view of the heart, which is the view that a physician observes during open heart surgery. Therefore, the left heart pump is on the right hand side of the FIG. 1 and the right heart pump is on the left hand side of FIG. 1. Each heart pump includes an upper chamber called an atrium and a lower chamber called a ventricle. The left heart pump therefore contains a left atrium (LA) and a left ventricle (LV), separated by a valve called the mitral valve. The right heart pump contains a right atrium (RA) and a right ventricle (RV), separated by a valve called the tricuspid valve.

The blood flows in the circulatory system in the following path: from the peripheral venous system (blood which has transferred through the body organs) to the RA, from the RA to the RV through the tricuspid valve, from RV to the pulmonary artery through the pulmonary valve, to the lungs. Oxygenated blood from the lungs is drawn from the pulmonary vein to the LA, from the LA to the LV through the mitral valve, and finally, from the LV to the peripheral arterial system (transferring blood to the organs of the body) through the aortic valve.

Normally, the heart pumps operate in synchrony and ensure the proper pumping action to provide oxygenated blood from the lungs to the organs of the body. A normal heart provides this synchrony by a complex conduction system which propagates electrical pulses to the heart muscle tissue to perform the necessary atrial and ventricular contractions. A heartbeat is the result of a regular train of electrical pulses to the proper portions of the heart to provide rhythmic heart pumping. The heart muscle provides pumping by the contraction of muscle tissue upon receipt of an

electrical signal, and the pumping action is made possible through a system of heart valves which enable blood flow in a single direction. Thus, the heart includes a complex electrical and mechanical network.

To pump blood through the circulatory system, a beating heart performs a cardiac cycle. A cardiac cycle consists of a systolic phase and a diastolic phase. During systole, the ventricular muscle cells contract to pump blood through both the pulmonary circulation and the systemic circulation. During diastole, the ventricular muscle cells relax, which causes pressure in the ventricles to fall below that in the atria, and the ventricles begin to be refilled with blood.

In normal condition, the cardiac pumping is highly efficient. One aspect of this high efficiency is due to sequential atrio-ventricular contraction. Near the end of diastole, the atria contract, causing an extra amount of blood to be forced into the ventricles. Thus, the ventricles have more blood (preload) to pump out during next systole. Another aspect of this high efficiency in blood pumping is contributed from a network or fast ventricular conduction system. As shown in FIG. 1, the system includes right and left bundle branches of conductive tissues that extend from the Bundle of His and the massive network of fast conducting Purkinje fibers that cover most of the endocardial surface of the ventricles. Electrical signals coming from the atrium are relayed to the Purkinje fibers through the bundle branches, and to the different regions of the ventricles by the Purkinje fiber network. Therefore the entire ventricular muscle cells can contract synchronously during systole. This synchronized contraction enhances the strength of the pumping power.

To assess the cardiac function, it is important to examine the LV systolic performance which directly determines the ability of the heart to pump blood through the systemic circulation. There are multiple ways to assess the performance of the heart. One way is to examine how well the LV contracts in order to determine the effectiveness of the LV as a pump. As can be seen from FIG. 2, the LV starts to contract after an electrical signal propagating down the left bundle branches stimulates

muscle cells of septal wall M and lateral wall N. In FIG. 3, the walls M and N are contracting such that they are forced towards each other to pump blood out of the ventricle. One measure of LV contraction effectiveness is called "contractility." Left ventricular contractility is a measure of overall strength of the contracting power of the LV muscle cells. It is a function of the health of the LV muscle tissue and the coordination of the contractions of the entire LV, including walls M and N. Such coordination depends on the health of the left bundle branches and on the health of the fast conducting Purkinje fiber network. LV contractility is estimated by measuring the peak positive rate of change of the LV pressure during systole. In mathematical terms, this is the maximum positive derivative of the LV pressure, which is denoted by the term "LV +dp/dt".

LV systolic performance is also measured by stroke volume, which is the volume of blood pumped out of the LV per systole. Stroke volume can be estimated by measuring aortic pulse pressure (PP).

Cardiac muscle cells need to be electrically excited before they can have a mechanical contraction. During the excitation (depolarization), electrical signals will be generated and they can be recorded both intracardially and extracardially. The recorded signals are generally called electrocardiogram (ECG). An ECG recorded intracardially is also called an electrogram, which is recorded from an electrode placed endocardially or epicardially in an atrium or a ventricle. An ECG recorded extracardially is often called surface ECG, because it is usually recorded from two or more electrodes attached to the skin of the body. A complete surface ECG recording is from 12-lead configuration.

The features in ECG are labeled according to the origin of the electrical activity. The signals corresponding to intrinsic depolarization in an atrium and a ventricle are called P-wave and QRS complex, respectively. The QRS complex itself consists of a Q-wave, a R-wave, and a S-wave. The time interval from P-wave to R-wave is called PR interval. It is a measure of the delay between the electrical excitation in the atrium

and in the ventricle.

Several disorders of the heart have been studied which prevent the heart from operating normally. One such disorder is from degeneration of the LV conduction system, which blocks the propagation of electric signals through some or all of the fast conducting Purkinje fiber network. Portions of the LV that do not receive exciting signals through the fast conducting Purkinje fiber network can only be excited through muscle tissue conduction, which is slow and in sequential manner. As a result, the contraction of these portions of the LV occurs in stages, rather than synchronously. For example, if the wall N is affected by the conduction disorder, then it contracts later than the wall M which is activated through normal conduction. Such asynchronous contraction of the LV walls degrades the contractility (pumping power) of the LV and reduces the LV+dp/dt (maximum positive derivative of the LV pressure) as well.

Another disorder of the heart is when blood in the LV flows back into the LA, resulting in reduced stroke volume and cardiac output. This disorder is called mitral regurgitation and can be caused by an insufficiency of the mitral valve, a dialated heart chamber, or an abnormal relationship between LV pressure and LA pressure. The amount of the back flow is a complex function of the condition of the mitral valve, the pressure in the LV and in the LA, and the rate of blood flow through the left heart pump.

These disorders may be found separately or in combination in patients. For example, both disorders are found in patients exhibiting congestive heart failure (CHF). Congestive heart failure (CHF) is a disorder of the cardiovascular system. Generally, CHF refers to a cardiovascular condition in which abnormal circulatory congestion exists as a result of heart failure. Circulatory congestion is a state in which there is an increase in blood volume in the heart but a decrease in the stroke volume. Reduced cardiac output can be due to several disorders, including mitral regurgitation (a back flow of blood from the LV to the LA) and intrinsic ventricular conduction disorder (asynchronous contraction of the ventricular muscle cells), which are the two common

abnormalities among CHF patients.

Patients having cardiac disorders may receive benefits from cardiac pacing. For example, a pacing system may offer a pacing which improves LV contractility, (positive LV pressure change during systole), or stroke volume (aortic pulse pressure), however, known systems require complicated measurements and fail to provide automatic optimization of these cardiac performance parameters. Furthermore, the measurements are patient-specific and require substantial monitoring and calibration for operation. Therefore, there is a need in the art for a system which may be easily adapted for optimizing various cardiac parameters, including, but not limited to, LV contractility, (peak positive LV pressure change during systole, LV+dp/dt), and cardiac stroke volume (pulse pressure). The system should be easy to program and operate using straightforward patient-specific measurements.

Summary of the Invention

This patent application describes multiple ways to provide optimized timing for ventricular pacing by determining certain electrical or mechanical events in the atria or ventricles that have a predictable timing relationship to the delivery of optimally timed ventricular pacing that maximizes ventricular performance. This relationship allows prediction of an atrio-ventricular delay used in delivery of a ventricular pacing pulse relative to a contraction of the atrium to establish the optimal pacing timing. Also provided are embodiments for measuring these events and deriving the timing relationship above. Those skilled in the art will understand upon reading the description that other events may be used without departing from the present invention.

In several embodiments, these measurements are used to optimize ventricular contractility as measured by maximum rate of pressure change during systole. In other embodiments, these measurements are used to optimize stroke volume as measured by aortic pulse pressure. In other embodiments, a compromise timing of pacing is available to provide nearly optimal improvements in both peak positive pressure

change during systole and aortic pulse pressure. In one embodiment, this pacing is provided by adjusting the atrio-ventricular delay time interval, which is the time interval after an atrial contraction, to deliver a pacing pulse to achieve the desired cardiac parameter optimization.

In one embodiment, a look-up table relating time intervals between measured events to atrio-ventricular delay time intervals is provided for programming an implanted device for delivering pacing pulses. The look-up table is produced using any one of the multiple ways to provide optimized timing for ventricular pacing.

This summary of the invention is intended not to limit the claimed subject matter, and the scope of the invention is defined by attached claims and their equivalents.

Brief Description of the Drawings

FIG. 1 is a diagram of a heart showing the chambers and the nervous conduction system.

FIG. 2 is a diagram of a ventricle beginning contraction.

FIG. 3 is a diagram of a contracted ventricle.

FIG. 4A is a graph of left ventricle intrinsic pressure as a function of time as referenced to an intrinsic P-wave event.

FIG. 4B is a graph of left ventricle intrinsic electrogram as a function of time as referenced to an intrinsic P-wave event.

FIG. 4C is a timing diagram showing a marker of an intrinsic P-wave and the marker of a ventricular pacing pulse that is optimally timed for maximum LV contractility as referenced to a paced P-wave event;

FIG. 4D is a graph of left atrial intrinsic pressure as a function of time as referenced to an intrinsic P-wave event.

FIG. 4E is a timing diagram showing a marker of an intrinsic P-wave and the marker o f a ventricular pacing pulse that is optimally timed for maximum stroke

volume as referenced to a paced P-wave event.

FIG. 5 is a flow diagram for detection of a Q* event.

FIG. 6 shows embodiments of the apparatus for the present subject matter.

FIG. 7 shows an embodiment of a predetermined mapping according to the present subject matter.

FIG. 8 is a schematic illustration of an embodiment of portions of a cardiac rhythm management system and portions of an environment in which it is used

FIG. 9 is a schematic/block diagram illustrating one embodiment of portions of the cardiac rhythm management system of FIG. 8.

FIG. 10 is a block diagram illustrating a system for calculating atrioventricular delays.

FIG. 11 is an illustration of one embodiment of a look-up table for programming atrioventricular delays.

FIG. 12 is a flow chart showing one method for producing an atrioventricular delay look-up table.

Detailed Description

In the following detailed description, reference is made to the accompanying drawings which form a part hereof and in which is shown by way of illustration specific embodiments in which the invention can be practiced. These embodiments are described in sufficient detail to enable those skilled in the art to practice and use the invention, and it is to be understood that other embodiments may be utilized and that electrical, logical, and structural changes may be made without departing from the spirit and scope of the present invention. The following detailed description is, therefore, not to be taken in a limiting sense and the scope of the present invention is defined by the appended claims and their equivalents.

Some of the embodiments illustrated herein are demonstrated in an implantable cardiac pacemaker, which may include numerous pacing modes known in the art.

However, these embodiments are illustrative of some of the applications of the present system, and are not intended in an exhaustive or exclusive sense. For example, the present system is suitable for implementation in a variety of implantable and external devices.

The present system provides a means for optimizing cardiac systolic function based on different cardiac performance measurements. The present disclosure provides a number of embodiments useful for, among other things, optimizing cardiac pumping strength and stroke volume. The concepts described herein may be used in a variety of applications which will be readily appreciated by those skilled in the art upon reading and understanding this description. The cardiac performance measurements expressly provided herein include contractility, peak positive ventricular pressure change, stroke volume, and pulse pressure. Other cardiac performance may be maximized using the teachings provided herein, and therefore, the express teachings of this disclosure are not intended in an exclusive or limiting sense. These concepts are expressly described in terms of the left ventricle, however, applications to other chambers of the heart, including the right ventricle, may be readily appreciated by those skilled in the art without departing from the present invention.

The inventors of this subject matter performed numerous tests and experiments to develop a pacing system which may be used to treat cardiac disorders. The system includes method and apparatus which are useful for providing optimization of different cardiac performance parameters, including, but not limited to, ventricular contractility, maximum rate of pressure change during systole, stroke volume, and pulse pressure. The embodiments provided herein use right atrial (RA) sensing events to time the pacing of the left ventricle (LV), right ventricle (RV), or both (BV) to optimize cardiac performance parameters. However, it is understood that these teachings are applicable to other pacing configurations. The teachings herein provide, among other things, optimal pacing which is selectable for treating different cardiac disorders. The disorders include, but are not limited to, congestive heart failure (CHF), mitral

regurgitation, and ventricular conduction disorder. The optimal pacing taught herein includes embodiments which do not use patient-specific measurements of hemodynamic parameters, such as pressure, blood flow, or measurements not typically provided by implantable pacing devices, and the system is capable of automatic adjustment to meet the needs of a particular patient.

AVD Time Intervals

Implantable rhythm management devices such as pacemakers, are useful for treating patients with abnormal cardiac functions. One pacing therapy is called DDD pacing mode. In DDD pacing mode, pacing electrodes are placed in the atrium (for example, the RA) and one or both of the ventricles. These electrodes are also used to sense electric signals from the atrium and the ventricle(s). If the device senses a signal in the atrium, it will inhibit the delivery of a pacing pulse to the atrium, otherwise it will pace the atrium after the end of a predetermined time period. Whenever the device senses or paces the atrium, it generates an event marker and at the same time starts an atrio-ventricular delay (AVD) time interval. At the end of this delay interval, the device will pace the ventricle(s) if no signals from the ventricle(s) are sensed by the device. Systems which provide ventricular pacing signals relative to the P-wave of an electrocardiogram signal refer to atrio-ventricular time delay interval (AVD time interval) as the time delay from the sensed P-wave to the delivery of the ventricular pacing signal. In patients exhibiting ventricular conduction disorder, such as the CHF condition, therapy using an AVD time interval which is shorter than the PR time interval may provide improved contractility because patients with degeneration of their LV conduction system require pacing of the affected parts of the LV (for example, the lateral wall N) early enough so that the contraction may be in phase with other parts of the LV that are excited by intrinsic conduction (for example wall M). Properly timed ventricular pacing can make both walls M and N contract in phase for increased contractility.

Patients with decreased stroke volume benefit from a shorter AVD time interval to decrease the mitral regurgitation effects and increase aortic pulse pressure. In addition, for congestive heart failure (CHF) patients, their PR interval may be prolonged which reduces the AV synchrony to some extent. Such a reduction in AV synchrony may further increase mitral regurgitation, and reduce the effect of preload of the LV. Use of a shorter AVD time interval increases pulse pressure by forcing the contraction of the LV into an earlier period, thus reducing the effects of mitral regurgitation.

Optimization of Cardiac Ventricle Contractility and Maximum Left Ventricle Pressure Change during Systole

Left ventricle contractility (pumping power) and peak positive rate of change of left ventricle pressure during systole (abbreviated as "LV+dp/dt") are related cardiac performance parameters. For instance, increases in LV contractility are observed in measurements as increases in left ventricle pressure change during systole.

FIG. 4A shows an intrinsic or unpaced left ventricle pressure curve following a P-wave. The Y event is the onset of intrinsic LV pressure increase. FIG. 4B shows an intrinsic left ventricular electrogram which is a QRS complex following a P-wave. Q* is an electrical signal which occurs at the beginning of a QRS complex. R is the largest peak of the QRS complex. In FIG. 4B, the Q* event leads the Y event of FIG. 4A. FIG. 4C shows a timing diagram under an optimally paced condition in which the LV contractility is maximized. The AVD_c time interval is equal to the time between the P-wave marker and the ventricular pacing marker V and that pacing provides maximum LV contractility. It is therefore called an optimal atrio-ventricular delay for contractility. It is noted that in the FIG. 4C the P_p marker is from a paced condition, as opposed to the P_I markers in FIGS. 4A and 4B, which arise from intrinsic heart activity. Therefore P_p occurs at a different time than P_I. Additionally, the diagrams are not to scale.

In their experimentation, the inventors learned that when pacing for maximum contractility the Q*, Y, and R events had a relatively predictable timing relationship with respect to the V pacing signal that is optimally timed by AVD_c. Furthermore, the inventors learned that linear models could be created which map the PQ* interval (the time difference between a P event and a Q* event) to an optimal atrio-ventricular delay for maximum contractility, AVD_c (FIG. 7 showing one embodiment). Additionally, linear mappings are possible for PY and PR to AVD_c, however, each mapping may result in different coefficients.

In one embodiment, an intrinsic PQ* time interval is measured for a patient. This is the time interval between the P-wave and a Q* event when no pacing signal is applied. After the PQ* time interval is recorded and averaged, then a pacing signal is applied with varying atrio-ventricular delays while monitoring LV+dp/dt (peak positive left ventricular pressure change). Then the atrio-ventricular delay which produced the maximum LV+dp/dt (optimal contractility) is determined and named as AVD_c, and is paired with that patient's PQ* time interval. The PQ*, AVD_c pairs are generated for a number of other patients and the data are plotted. In one embodiment, a linear regression method is applied to determine a straight line approximation for AVD_c as a function of PQ*. The equation is: $AVD_c = K1(PQ^*) - K2$. A programmable device which measures the intrinsic PQ* interval can estimate AVD_c using this equation. Therefore, once K1 and K2 are determined, the calibration of the device is complete. This means that subsequent patients may have optimal contractility pacing without requiring the pressure measurements and additional calibration stages. As described below, the same procedures may be used with PY or PR, however, as stated before, the coefficients may be different.

This means that, if PQ* is measured, then a patient may receive optimal contractility pacing of the left ventricle using measurements of the P-wave and of Q*. In the case where PY is used instead of PQ*, then the measurements will be of the P-wave and of the Y event, which is the onset of pressure increase in the left ventricular

contraction. If the PR interval is used, then the measurements will be the P-wave and the R-wave of the QRS complex.

Therefore, given a patient's intrinsic PQ* or PY or PR time interval and the respective mapping, an AVD_c is calculated. This AVD_c is an approximation of the actual AVD_c using the mapping method.

It is noted that any event which is relatively constant with respect to the optimally timed V pacing signal (pacing using AVD_c) may be used as a predictable event for use in the present system. In one embodiment, an event which is relatively constant is one which has a deviation between the lesser of 20 ms or 25 percent of the population mean. Therefore, other embodiments incorporating events not expressly mentioned herein may be used without departing from the present system.

P-Wave Signal

When the electronic P-wave signal is used as a reference for any of the embodiments, the P-wave signal is detectable using devices including, but not limited to, catheters or external probes to create electrocardiograms. In one embodiment, the P-wave is sensed from the right atrium and used as a reference for the time interval measurements and pacing delivery. In some cases where a patient's atrium is paced then the P-wave pacing marker is used instead of the intrinsic P-wave.

PQ* Measurement and Mapping

As stated above, the inventors determined some "events" would have a predictable relationship to the optimally timed ventricular pacing signal. The Q* event was defined as one candidate because it is relatively constant relative to the LV pacing mark, V, at optimal timing for maximum contractility. Q* is an electrical signal which occurs at the beginning of a QRS complex. Therefore, in one embodiment of the system, the time delay between the P-wave and the Q* event is used to provide the linear variable to calculate AVD_c. In this embodiment, the equation is: $AVD_c = K1$

(PQ*) - K2.

Furthermore, the inventors of the present system realized that the PQ* interval provides a linear variable which may be used to estimate AVD_c using a single calibration procedure for determining the constants K1 and K2. One type of calibration was discussed above, mapping AVD_c, PQ* pairs in a linear fashion to provide K1 and K2. The PQ* and AVD_c information is then plotted on a two-dimensional chart and a linear regression method is performed to provide a best line fit through the sample point pairs. This linear fit provides the K1 and K2 coefficients.

In one study using 13 patients, an equation for AVD_c was generated which provided K1 equal to 0.94 and K2 equal to 55.7 milliseconds. In this equation, PQ* is measured in milliseconds. This equation is expressed as: $AVD_c = 0.94\ PQ^* - 55.7$ milliseconds. It is noted that the coefficients may vary and that estimated AVD_c may depart from the actual optimum AVD_c by approximately 20 percent and continue to provide near optimal performance within 80 percent of the maximum contractility. Furthermore, the coefficients may vary slightly depending on the number of samples taken in the calibration stage. Therefore, the coefficients provided herein may vary without departing from the present invention.

In one embodiment, the P-wave was detected using a threshold detector which indicated a P-wave at approximately 20 percent of the maximum P-wave amplitude in the right atrium. In one embodiment shown in FIG. 5, the Q* event is determined by passing the QRS complex as sampled from the left ventricle through a 5 point low-pass digital filter having a sampling time of 2 milliseconds, detecting the Q portion of the wave, calculating a maximum absolute value of the slope for the Q-wave, and indicating a point on the filtered Q-wave where the absolute value of the slope equals 2% of the absolute value slope of the Q-wave. Those skilled in the art will readily recognize that other determination methods may be used for P and Q* which do not depart from the present system. Changes in the measurement techniques and slope criteria do not depart from the present system.

In another embodiment, the coefficient of PQ*, K1, is assumed to be unity, and the coefficient K2 amounts to an offset time delay from the PQ* interval to predict or estimate the optimal AVD_c. In this embodiment, PQ* and AVD_c are sampled for a variety of patients at a variety of PQ* intervals and a variety of AVD_c to generate a mean offset time delay K2 for a number of patients. In this embodiment, the equation is as follows: AVD_c estimated = PQ* - Wa milliseconds. Using the previous data for the 13 patients, the equation is: AVD_c estimated = PQ* - 67 milliseconds. This embodiment provides an easier calculation, since a subtraction is less processor intensive than multiplications using floating point numbers. However, some accuracy is lost for the approximation.

It is noted that the coefficients may vary and that the estimated AVD_c may depart from the actual optimum AVD_c by approximately 20 percent and continue to provide near optimal performance within 80 percent of the maximum contractility. Furthermore, the coefficients may vary slightly depending on the number of samples taken in the calibration stage. Therefore, the coefficients provided herein may vary without departing from the present invention.

Those skilled in the art will readily recognize that other methods may be employed to generate other fits to the data which do not depart from the scope of the present invention.

In one embodiment, the measurements of the P-wave and Q* are provided using an electrode implanted in the right atrium and an electrode implanted in the left ventricle. A programmable pulse generator is used to sense the P-wave and measure the time between occurrence of a sensed P-wave and a sensed Q* event. The Q* event is determined by electronics in the pulse generator which perform the required slope and comparison operations to determine Q*. After a PQ* time interval is determined, the AVD_c is determined using any of the embodiments described herein and their equivalents. Once the AVD_c is determined, it may be used in the next pacing interval to provide an optimized atrio-ventricular delay based on the PQ* time interval.

It is understood that the Q* event may be defined differently and provide substantially the same results with a different set of parameters, K1 and K2. Furthermore, any electrical signal event which bears a predictable relationship to the beginning of intrinsic LV electrogram signals may be used in place of Q*. For example, in one embodiment the beginning of the RV electrogram may be used in place of Q*. Or in another embodiment, the Q* may be measured by surface ECG as the onset of the signal averaged QRS complex. Furthermore, information from more than one lead may be used to more accurately determine Q*.

PR Measurement and Mapping

In another embodiment, the R-wave peak, which is the largest peak of the QRS complex of an intrinsic LV electrogram, is used since it has a predictable relationship to the delivery of optimally timed ventricular pacing for maximum contractility. In particular, the linear time relationship may be derived in terms of the PR interval for optimal atrio-ventricular delay for optimal left ventricular pressure change during systole. In this case, the equation is: $AVD_c = N1\ PR - N2$, where AVD_c is for pacing the LV, and PR is the time interval from right atrial sensing marker to the largest peak of the QRS complex of intrinsic LV electrogram. In one embodiment, the N1 and N2 coefficients are determined by mapping the PR time interval to the optimal AVD_c for a number of patients for optimal left ventricular pressure change during systole. In one study using 13 patients, the coefficient N1 is equal to 0.82 and the coefficient N2 is equal to 112 milliseconds. The equation for this calibration is: $AVD_c = 0.82\ PR - 112$ milliseconds. It is noted that the coefficients may vary and that the estimated AVD_c may depart from the actual optimum AVD_c by approximately 20 percent and continue to provide near optimal performance within 80 percent of the maximum contractility. Furthermore, the coefficients may vary slightly depending on the number of samples taken in the calibration stage. Therefore, the coefficients provided herein may vary without departing from the present invention.

In another embodiment, the N1 coefficient is assumed to be unity, and the PR, AVD_c data pairs are averaged to provide a linear dependence with an offset equal to N2. This embodiment provides an easier calculation, since a subtraction is less processor intensive than multiplications using floating point numbers. However, some accuracy is lost for the approximation. For example, using data in the previous study: $AVD_c = PR - 159$ milliseconds. In one embodiment, the R-wave signal is measured by detecting the largest peak of the QRS complex of the intrinsic LV electrogram. Therefore, electrical signals are used in this embodiment to provide the PR time interval, and therefore the optimal atrio-ventricular delay for optimal left ventricular pressure change during systole. The coefficients N1 and N2 are provided in an initial calibration stage, which means that subsequent readings using this embodiment generate the optimal AVD_c automatically upon detection of the PR time interval. Furthermore, the N1 and N2 variables may change in value without departing from the teachings provided herein.

Other features of the QRS complex may be used for measurement. As stated above, these events may be used as long as they have a predictable timing relationship to the delivered pacing for optimal contractility. It is noted that the coefficients may vary and that the estimated AVD_c may depart from the actual optimum AVD_c by approximately 20 percent and continue to provide near optimal performance within 80 percent of the maximum contractility. Furthermore, the coefficients may vary slightly depending on the number of samples taken in the calibration stage. Therefore, the coefficients provided herein may vary without departing from the present invention.

PY Measurements and Mappings

In another embodiment, a mechanical event is provided as a reference instead of an electrical event. In one embodiment, the mechanical event, Y is determined as the beginning of intrinsic LV pressure development. This means that a pressure transducer such as a micromonometer can provide instantaneous pressure data in the left ventricle.

In this embodiment, the atrio-ventricular delay optimized for maximum left ventricular pressure change during systole is provided as: $AVD_c = M1\ PY - M2$. In one embodiment, a micromonometer is placed in the LV to measure left ventricular pressure change during systole. The PY time interval, which is the time interval from right atrial sensing of the P-wave to the beginning of the intrinsic LV pressure development, is mapped to recorded AVD_c values for maximum left ventricular pressure change during systole. This mapping is plotted to perform a linear regression in order to determine the coefficients M1 and M2. In one study, M1 is equal to 0.96 and M2 is equal to 139 milliseconds. Therefore, in this study, the $AVD_c = 0.96\ PY - 139$ milliseconds. It is noted that the coefficients may vary and that the estimated AVD_c may depart from the actual optimum AVD_c by approximately 20 percent and continue to provide near optimal performance within 80 percent of the maximum contractility. Furthermore, the coefficients may vary slightly depending on the number of samples taken in the calibration stage. Therefore, the coefficients provided herein may vary without departing from the present invention.

In another embodiment, the M1 coefficient is approximated as unity, and then the PY and AVD_c pairs are used to determine a linearized mapping which amounts to: $AVD_c = PY - N_a$, where N_a is an averaged offset delay for the samples taken. In one embodiment, $AVD_c = PY - 150$ milliseconds. This embodiment provides an easier calculation, since a subtraction is less processor intensive than multiplications using floating point numbers. However, some accuracy is lost for the approximation. Again, it is noted that the coefficients may vary and that the estimated AVD_c may depart from the actual optimum AVD_c by approximately 20 percent and continue to provide near optimal performance within 80 percent of the maximum contractility. Furthermore, the coefficients may vary slightly depending on the number of samples taken in the calibration stage. Therefore, the coefficients provided herein may vary without departing from the present invention.

Other mechanical events may be used as long as they are relatively predictable

with respect to the Y event. The Y events may be selected from signals including, but not limited to, ventricular pressure, cardiac phonogram, cardiac acoustic signals (such as recorded from an accelerometer external to or inside an implantable device), Doppler recording of atrio-ventricular valve motion, and M-mode, 2D, or 3D echo imaging of ventricular wall motion (FIG. 6 showing one embodiment of the mechanical event sensor).

Stroke Volume Optimization Using Atrio-Ventricular Delay

Stroke volume is related to pulse pressure. The inventors discovered that for maximum pulse pressure (stroke volume), there is a predictable timing relationship between an optimally delivered ventricular pulse V and the peak of left atrial systole, X. Therefore, the optimal atrio-ventricular delay for maximum pulse pressure, AVD_s, is determined by PX time interval measurements, as shown in FIG. 4E.

In one embodiment, stroke volume is optimized by determining the atrio-ventricular delay for maximum aortic pulse pressure, AVD_s. In one embodiment, the X event is measured by placing a pressure sensing catheter inside the LA. In another embodiment, the X event is detected by measuring the LV pressure, because the LA contraction is seen in the LV pressure curve by a pre-systolic component. The peak of the LA systole is considered the same as the pre-systolic pressure in the LV pressure curve. The time interval between P and the pre-systolic component of LV pressure provides a linear equation. Therefore, in order to generate the linear mapping of PX to AVD_s, a number of PX, AVD_s pairs are generated by measuring maximum aortic pulse pressure for varying PX. The linear relationship is expressed by: $AVD_s = M3\ PX - M4$ milliseconds. In one embodiment, a calibration procedure was performed to generate a number of PX, AVD_s pairs, which are mapped and a best line fit is performed to determine M3 and M4. In one embodiment, M1 is equal to 1.22 and M2 is equal to 132 milliseconds. Therefore, the AVD_s relationship is: $AVD_s = 1.22\ PX - 132$ milliseconds. It is noted that the coefficients may vary and that the estimated

AVD_s may depart from the actual optimum AVD_s by approximately 20 percent and continue to provide near optimal performance of the maximum stroke volume. Furthermore, the coefficients may vary slightly depending on the number of samples taken in the calibration stage. Therefore, the coefficients provided herein may vary without departing from the present invention.

In one embodiment, the P-wave event is measured using a threshold detection where the P-wave is determined to be 20 % of the maximum P-wave amplitude. Other detection methods for the P-wave may be used without departing from the present system. The X event may be determined by several ways, including but not limited to: locating the point of maximum atrial pressure, Doppler measurements, and S4 components of accelerator measurements.

Other embodiments using different values for M3 and M4 are possible without departing from the present system. Furthermore, other markers may be used which are directly related to the PX time interval provided in one embodiment.

It is noted that any event which is relatively constant with respect to the optimally timed V pacing signal (pacing using AVD_s) may be used as a predictable event for use in the present system. In one embodiment, an event which is relatively constant is one which has a deviation between the lesser of 20 ms or 25 percent of the population mean. Therefore, other embodiments incorporating events not expressly mentioned herein may be used without departing from the present system.

Selection of Atrio-Ventricular Delay for Improved Contractility and Stroke Volume

Depending on the condition of a heart and its disorders, optimal atrio-ventricular delay for maximum contractility may provide especially nonoptimal stroke volume. Likewise, optimal atrio-ventricular delay for maximized stroke volume may result in nonoptimal contractility. Therefore, in order to provide a compromised atrio-ventricular delay which provides an approximately optimal atrio-ventricular delay for

both contractility and stroke volume, AVD_{cs}, it is desirable to have an atrio-ventricular delay which provides near optimal contractility and near optimal stroke volume. The inventors of the present system derived a relationship which provides a compromise between optimal contractility and optimal stroke volume. In one embodiment, the optimized atrio-ventricular delay, AVD_{cs}, is a linear relationship in the PR time interval, as follows: $AVD_{cs} = K3\ PR_m - K4$ milliseconds. PR_m is a time interval measured from a right atrial sensing marker, P, to a right ventricular sensing marker, R_m. In one embodiment, the compromised AVD_{cs} is provided by determining AVD_c and AVD_s for a number of PR values and for a number of patients. Then a linear regression provides a best line fit for both contractility and stroke volume. In one embodiment, AVD_{cs} equals $0.5\ PR_m - 15$ milliseconds, where AVD_{cs} is for pacing at least one ventricle, and where the time interval PR_m is measured from a right atrial sensing marker, P, to a right ventricular sensing marker, R_m. In this embodiment, the resulting atrio-ventricular delay provides a left ventricular pressure change within 90% of the optimal left ventricular pressure change during systole. Furthermore, this embodiment provides an aortic pulse pressure which is within 80% of the optimal aortic pulse pressure. It is noted that the coefficients may vary and still provide a reasonable approximation of AVD_{cs}. For example, in one embodiment K3 may be in the range from 0.4 to 0.6 and K2 may be in the range from 0 to 30 ms. Therefore, the present system offers flexibility in the selection of coefficients, and those provided are demonstrative and not an exclusive set of coefficients.

In one embodiment, a left ventricular event is used to provide a time interval for calculation of AVD_{cs}. In one case the LV event is the LV R-wave. The LV R-wave marker signal may also be used as an event. It is noted that any event which is relatively constant with respect to the near optimally timed V pacing signal may be used as a predictable event for use in the present system. In one embodiment, an event which is relatively constant is one which has a deviation between the lesser of 20 ms or 25 percent of the population mean. Therefore, other embodiments incorporating events

not expressly mentioned herein may be used without departing from the present system.

In one embodiment, the left ventricular R wave is used to develop a relationship between the PR interval (the time interval between a P event and an R event) and AVD_{cs}. For a particular patient, the intrinsic PR interval is measured. Additionally, a sweep of atrio-ventricular delays are applied to the pacing of the patient and LV+dp/dt and pulse pressure are measured for each different atrio-ventricular delay. The LV+dp/dt data is plotted against a normalized value of the atrio-ventricular delay. Additionally, the pulse pressure is also plotted against a normalized value of the atrio-ventricular delay. In one embodiment, the atrio-ventricular delay is divided by PR-30 ms to normalize the delay. The tests are performed for a number of additional patients and the normalized plots are mapped. Then an averaging of the various LV+dp/dt vs. normalized atrio-ventricular delay data is performed. An averaging of the pulse pressure data vs. normalized atrio-ventricular delay data is also performed. The atrio-ventricular delay (normalized value) at the LV+dp/dt curve peak is used as an optimal averaged atrio-ventricular delay. The peak of the pulse pressure curve is also determined. In one example, the optimal averaged normalized atrio-ventricular delays for both curves was determined to be approximately 0.50 times the normalized PR time interval, or 0.50(PR-30) milliseconds.

In one study data was taken using a series of intermittent pacing (5 pacing beats in every 15 sinus beats) from one of three sites (RV, LV, and BV) at one of five AV delays (equally spaced between 0 msec and PR-30 msec). Each pacing site/AV delay combination was repeated five times in random order. Pressure and electrogram data were recorded from the ventricles. LV+dp/dt and PP were measured from LV and aortic pressure recordings on a beat-by-beat basis. For each paced beat, values of the LV+dp/dt and PP were compared to a preceding 6-beats-averaged sinus baseline. Then the response to pacing configuration was averaged. However, other measurements may be taken to obtain the required information.

Switchable Pacing Therapies

Any of the teachings provided herein may be employed in a variety of cardiac devices, including implantable pacing devices. In one embodiment, an implantable device also includes means for changing the ventricular pacing to adjust for maximum contractility, maximum stroke volume or a compromise providing nearly optimal contractility and stroke volume. In such an embodiment, the pacing system contemplates the use of all of the different optimal atrio-ventricular delays to adjust the therapy to a cardiac patient. In one embodiment AVD_{cs} is used as a default atrio-ventricular pacing delay, which may be maintained or modified at a later time depending on the therapy required. For example, in one embodiment of the system, the pacing initiates with an atrio-ventricular delay equal to AVD_{cs}. If at any time an optimal contractility is required, the atrio-ventricular pace delay is changed to AVD_{c}. Additionally, if at any time optimal stroke volume is required, the atrio-ventricular delay is changed to AVD_{s}. Other variations and combinations are possible without departing from the present invention. Furthermore, the switching of the pacing therapies may be provided by an external instruction, such as a programmer, or by an internally executing software for selecting the appropriate therapy. Other ways of switching between therapies may be encountered which do not depart from the present system.

AVD Time Intervals in a DDD Pacing Mode

In one embodiment, a DDD pacing mode requires a post-sensing AVD and a post-pacing AVD. The post-sensing AVD is applied following an intrinsic, i.e., sensed, P-wave. The post-pacing AVD is applied following a paced P-wave, or a delivery of pacing pulse to the atrium. When intracardiac electrogram is used for P-wave detection, the atrial event that starts an AVD is referred to as an A event. Accordingly, an intrinsic atrial contraction (sensed P-wave) is referred to as a sensed A event, and a paced atrial contraction (paced P-wave) or a delivery of atrial pacing pulse is referred to

a paced A event. When intracardiac electrogram is used for R-wave detection, a ventricular event is referred to as a V event. Accordingly, an intrinsic ventricular contraction is referred to as a sensed V event; a paced ventricular contraction or a delivery of ventricular pacing pulse is referred to a paced V event.

All of the methods for calculating AVD based on PR, PQ*, PX, and PY time intervals as discussed above in this document apply regardless of whether P represents a sensed A event or a paced A event. In one application, a delivery of atrial pacing pulse is used, instead of a paced atrial contraction or a paced P-wave, for AVD and other timing purposes. Thus, the PR, PQ*, PX, and PY time intervals discussed above are generalized as AV, AQ*, AX, and AY time intervals, respectively, with A referring to either a sensed A event or a paced A event.

Thus, post-sensing AVD time intervals and post-pacing AVD time intervals are calculated using the formulas discussed above with AV, AQ*, AX, and AY time intervals substituting for PR, PQ*, PX, and PY time intervals, respectively. A post-sensing AVD is calculated using one of the formulas discussed above with post-sensing AV, AQ*, AX, and AY time intervals, i.e., AV, AQ*, AX, and AY time intervals measured with a sensed A event, respectively. A post-pacing AVD is calculated using one of the formulas discussed above with post-pacing AV, AQ*, AX, and AY time intervals, i.e., AV, AQ*, AX, and AY time intervals measured with a paced A event, respectively. For purpose of discussion, the A, V, Q*, X, and Y events for measuring post-sensing AV, AQ*, AX, and AY time intervals are hereinafter referred to as post-sensing A, V, Q*, X, and Y events, respectively, and the A, V, Q*, X, and Y events for measuring post-pacing AV, AQ*, AX, and AY time intervals are hereinafter referred to as post-pacing A, V, Q*, X, and Y events, respectively.

The AV, AQ*, AX, and AY time intervals are measured between an A event and the V, Q*, X, and Y events, respectively, that are subsequent and closest to the A event. In one embodiment, V events includes right ventricular events. In another embodiment, V events includes left ventricular events.

Pacing System with Adjustable AVD Time Intervals

FIG. 8 is a schematic illustration of an embodiment of portions of a cardiac rhythm management system 800 and portions of an environment in which it is used. System 800 includes a dual-site or multi-site pacing system capable of performing DDD mode pacing with one or more adjustable AVD time intervals calculated by using one or more of the formulas discussed in this document. In one embodiment, system 800 is a cardiac rhythm management system including, among other things, an implanted device 810 and an external programmer 840. Implanted device 810 is implanted within a patient's body 801 and coupled to the patient's heart 802 by a lead system 805. Examples of implanted device 810 include pacemakers, pacemaker/defibrillators, and cardiac resynchronization therapy (CRT) devices. Programmer 840 includes a user interface for system 800. A "user" refers to a physician or other caregiver who examines and/or treats the patient with system 800. The user interface allows a user to interact with implanted device 810 through a telemetry link 870.

In one embodiment, as illustrated in FIG. 8, telemetry link 870 is an inductive telemetry link supported by a mutual inductance between two closely-placed coils, one housed in a wand 875 near or attached onto body 801 and the other housed in implanted device 810. In an alternative embodiment, telemetry link 870 is a far-field telemetry link. In one embodiment, telemetry link 870 provides for data transmission from implanted device 810 to programmer 840. This may include, for example, transmitting real-time physiological data acquired by implanted device 810, extracting physiological data acquired by and stored in implanted device 810, extracting therapy history data stored in implanted device 810, and extracting data indicating an operational status of implanted device 810 (e.g., battery status and lead impedance). In one specific embodiment, the real-time or stored physiological data acquired by implanted device 810 includes one or more signals allowing for measurement of one or

more of the post-sensing and/or post-pacing AV, AQ*, AX, and AY time intervals, which are used to calculate the one or more adjustable AVD time intervals. In another specific embodiment, the real-time or stored physiological data acquired by implanted device 810 includes presentations, such as event markers, of one or more of the post-sensing and/or post-pacing A, V, Q*, X, and Y events, which are detected by implantable device 810. In yet another specific embodiment, the real-time or stored physiological data acquired by implanted device 810 includes one or more of the post-sensing and/or post-pacing AV, AQ*, AX, and AY time intervals, which are measured by implantable device 810. This allows programmer 840 to calculate the one or more adjustable AVD time intervals. In a further embodiment, telemetry link 870 provides for data transmission from programmer 840 to implanted device 810. This may include, for example, programming implanted device 810 to acquire physiological data, programming implanted device 810 to perform at least one self-diagnostic test (such as for a device operational status), and programming implanted device 810 to deliver at least one therapy. In one embodiment, programming implanted device 810 includes sending therapy parameters to implantable device 810. In one specific embodiment, the therapy parameters include the one or more adjustable AVD time intervals each calculated to provide for an approximately optimal hemodynamic performance. Depending on the conditions and needs a particular patient, the one or more adjustable AVD time intervals are calculated using one of the formulas provided in this document to approximately optimize the patient's contractility (i.e., ventricular synchrony) by maximizing the LV+dp/dt, and/or stroke volume by maximizing aortic pulse pressure, as discussed above.

FIG. 9 is a schematic/block diagram illustrating one embodiment of portions of system 800. System 800 includes an implanted portion and an external portion. The implanted portion resides within body 801 and includes implanted device 810 and lead system 805 providing for electrical connection between implanted device 810 and heart 802. The external portion includes programmer 840 and wand 875 connected to

programmer 840. Telemetry link 870 provides for bi-directional communications between implanted device 810 and programmer 840.

In one embodiment, lead system 805 includes one or more leads having endocardial electrodes for sensing cardiac signals referred to as intracardiac ECGs, or electrograms. In one embodiment, lead system 805 includes at least an atrial lead and a ventricular lead. In one embodiment, as illustrated in FIG. 9, lead system 805 includes an atrial lead 805A having at least one electrode placed within the right atrium, a right ventricular lead 805B having at least one electrode placed within the right ventricle, and a left ventricular lead 805C having at least one electrode placed in or about the left ventricle. In one specific embodiment, lead 805C includes at least one electrode placed in coronary venous vasculature traversing the left ventricle. Such lead system allows for CRT including left ventricular, right ventricular, or biventricular pacing.

In one embodiment, implanted device 810 includes a sensing circuit 921, a therapy circuit 922, an implant controller 923, an implant telemetry module 924, a coil 925, a mechanical event sensor 935, a sensor interface circuit 936, and a power source 920. Sensing circuit 921 includes sensing amplifiers each sense a cardiac signal from a cardiac location where an endocardial electrode of lead system 805 is placed. The cardiac signals are indicative of the post-sensing and post-pacing A, V, and Q* events. Therapy circuit 922 includes pacing output circuits each delivering pacing pulses to a cardiac location where an endocardial electrode of lead system 805 is placed. Mechanical event sensor 935 includes at least one of a sensor that senses a mechanical signal indicative an onset on intrinsic LV pressure increase, i.e. the Y event, and a sensor that senses a mechanical signal indicative of a peak of LV presystolic pressure, i.e., the X event. In one alternative embodiment, mechanical event sensor 935 includes a sensor that senses an end of left atrium systolic pressure as the X event. Sensor interface circuit 936 conditions each mechanical signal. Implant controller 923 controls the operation of implanted device 810. In one embodiment, implant controller 923 includes a memory circuit on which at least one therapy algorithm and therapy

parameters are stored. The controller executes the therapy instructions to deliver pacing pulses to heart 802 with the therapy parameters. In one embodiment, the therapy algorithm are programmed into the memory circuit when implant device 810 is built, and the therapy parameters are programmed into the memory circuit by programmer 840 via telemetry link 870. In another embodiment, both the therapy instructions and parameters are programmed to the memory circuit by programmer 840 via telemetry link 870. In one embodiment, the therapy parameters stored in the memory circuit are dynamically updated by programmer 840 via telemetry link 870 during or between therapy deliveries. In one embodiment, the therapy algorithm controls the pacing pulse delivery by using the therapy parameters calculated based on one or more of the cardiac signals and/or one or more of the mechanical signals. The therapy parameters including at least one AVD time interval. Implant controller 923 includes a therapy timing controller to time each delivery of ventricular pacing pulse according to the at least one AVD time interval.

Implant telemetry module 924 and coil 925 constitute portions of implanted device 810 that support telemetry link 870. Power source 920 supplies all energy needs of implanted device 910. In one embodiment, power source 920 includes a battery or a battery pack. In a further embodiment, power source 920 includes a power management circuit to minimize energy use by implant device 810 to maximize its life expectancy.

In one embodiment, programmer 840 includes a signal processor 950, a therapy controller 960, a display 941, a user input module 942, and a programmer telemetry module 945. Programmer telemetry module 945 and wand 875, which is electrically connected to programmer telemetry module 945, constitute portions of programmer 840 that support telemetry link 870. In one embodiment, signal processor 950 receives signals transmitted from implanted device 810 via telemetry link 870 and processes the signals for presentation on display 941 and/or use by therapy controller 960. In one embodiment, the received signals include the one or more of the cardiac signals,

representations of cardiac events such as the event markers, the one or more of the mechanical signals, and/or representations of the mechanical events. In another embodiment, the received signals include parameters measured from the one or more of the cardiac signals and/or the one or more of the mechanical signals. In one embodiment, therapy controller 960 generates therapy parameters to be transmitted to implanted device 810 via telemetry link 870. In one embodiment, therapy controller 960 receives user-programmable parameters from user input module 942 and converts them into code recognizable by implanted device 810. In one embodiment, therapy controller 960 calculates the one or more adjustable AVD time intervals based on the signals received from implanted device 810 via telemetry link 870. In one embodiment, user input module 942 receives commands from the user to control the data acquisition and/or pacing operations of implanted device 810. In one embodiment, display 941 is an interactive display that includes at least portions of user input module 942, such that the user may enter commands by contacting display 941.

In one embodiment, implant controller 923 receives the one or more of the cardiac signals and the one or more of the mechanical signals required for AVD calculation. Implant controller 923 detects one or more of the post-sensing and post-pacing A, V, Q*, X, and Y events from the one or more of the cardiac signals and the one or more of the mechanical signals, measures one or more of the post-sensing and post-pacing AV, AQ*, AX, and AY time intervals, and calculates the one or more adjustable AVD time intervals by using one or more of the formulas discussed in this document.

In another embodiment, implant controller 923 receives the one or more of the cardiac signals and the one or more of the mechanical signals. These signals are transmitted to programmer 840 through telemetry link 870. Programmer 840 receives the transmitted signals, from which it detects one or more of the post-sensing and post-pacing A, V, Q*, X, and Y events, measures one or more of the post-sensing and post-pacing AV, AQ*, AX, and AY time intervals, and calculates the one or more adjustable

AVD time intervals by using one or more of the formulas discussed in this document. Programmer 840 sends the calculated AVD time intervals to implanted device 810 to control the pacing pulse deliveries.

In yet another embodiment, implant controller 923 receives the one or more of the cardiac signals and the one or more of the mechanical signals, detects one or more of the post-sensing and post-pacing A, V, Q*, X, and Y events from these signals, and measures one or more of the post-sensing and post-pacing AV, AQ*, AX, and AY time intervals. The one or more of the post-sensing and post-pacing AV, AQ*, AX, and AY time intervals are transmitted to programmer 840 through telemetry link 870. Programmer 840 calculates the one or more adjustable AVD time intervals by using one or more of the formulas discussed in this document, and sends the calculated AVD time intervals to implanted device 810 to control the pacing pulse deliveries.

In still another embodiment, implant controller 923 receives the one or more of the cardiac signals and the one or more of the mechanical signals and detects one or more of the post-sensing and post-pacing A, V, Q*, X, and Y events from these signals. The representations, such as event markers, of the one or more of the post-sensing and post-pacing A, V, Q*, X, and Y events are transmitted to programmer 840 through telemetry link 870. Programmer 840 measures the one or more of the post-sensing and post-pacing AV, AQ*, AX, and AY time intervals based on the representations of the one or more of the post-sensing and post-pacing A, V, Q*, X, and Y events and calculates the one or more adjustable AVD time intervals by using one or more of the formulas discussed in this document, and sends the calculated AVD time intervals to implanted device 810 to control the pacing pulse deliveries.

The above embodiments for calculating the one or more adjustable AVD time intervals based on the one or more of the cardiac signals and/or mechanical signals are selected, modified, combined, and/or mixed in commercial embodiments, depending on, among other things, the objective of the pacing therapy and computational resources available in implanted device 810. Embodiments using programmer 840 to perform

detection, measurement, and calculation allow for implementation of AVD optimization systems without requiring an implanted device specifically configured for this purpose. In one embodiment, programmer 840 performs functions that implanted device 810 is not capable of performing. For example, a patient may have an implanted device capable of detecting and telemetering only A and V events. To optimize contractility, or ventricular synchrony, by using an approximately optimal AVD calculated based on AV, an external programmer measures the AV time interval and calculates the approximately optimal AVD. If the approximately optimal AVD is to be calculated from AQ*, the external programmer detects Q* from one of the cardiac signals telemetered from the implanted device, measures AQ*, and calculates the approximately optimal AVD.

In one embodiment, programmer 840 is a computer-based device. Signal processor 950 and therapy controller 960 are each implemented as one of a hardware, a firmware, a software, or a combination of any of these. In one embodiment, signal processor 950 and therapy controller 960 each include software that is to be installed on programmer 840 when the AVD calculation, or at least a portion of the AVD calculation, is intended to be performed with that programmer. In one embodiment, the software supporting the AVD calculation is stored on one or more storage media that allow for installation when needed.

FIG. 10 is a block diagram illustrating a system 1080 for calculating AVD using one or more of the methods discussed above. System 1080 includes a signal input 1081, an event detector 1083, a measurement module 1085, an AVD calculator 1087, and a memory circuit 1089. In one embodiment, system 1080 is included in implanted device 810, such as being implemented as part of implant controller 923. In another embodiment, system 1080 is included in programmer 840, such as being implemented as part of signal processor 950 and therapy controller 960. In yet another embodiment, portions of system 1080 are included in implanted device 810 and programmer 840. In

other words, systems 1080 includes portions of both implanted device 810 and programmer 840.

Signal input 1081 receives signals from a signal source including at least one of sensing circuit 921 and mechanical event sensor 935. In one embodiment, Signal input 1081 includes a cardiac signal input that receives the one or more of the cardiac signals sensed by sensing circuit 921. In one embodiment, signal input 1081 further includes a mechanical signal input that receives the one or more of the mechanical signals sensed by mechanical event sensor 935. In one embodiment, signal input 1081 is included in implanted device 810 and receives the signals from the signal source without using telemetry link 870. In another embodiment, signal input 1081 is included in programmer 840 and receives the signals from the signal source via telemetry link 870. The one or more of the cardiac signal are indicative of one or more of post-sensing and post-pacing A, V and Q* events. The one or more of the mechanical signals are indicative of one or more of the post-sensing and post-pacing X and Y events. Event detector 1083 detects the events required for AVD calculation, including one or more of the post-sensing and post-pacing A, V, Q*, X, and Y events. Measurement module 1085 measures time intervals between two of these events. The time intervals include at least one of the post-sensing AV, AQ*, AX, and AY time intervals and post-pacing AV, AQ*, AX, and AY time intervals. In one embodiment, measurement module 1085 measures one or more of post-sensing AV, AQ*, AX, and AY time intervals. AVD calculator 1087 then calculates one or more post-sensing AVD time intervals based on the one or more of the post-sensing AV, AQ*, AX, and AY time intervals according to the formulas presented above. This is sufficient for a VDD mode pacing. In another embodiment, measurement module 1085 measures one or more of post-pacing AV, AQ*, AX, and AY time intervals in addition to the one or more of post-sensing AV, AQ*, AX, and AY time intervals. AVD calculator 1087 then calculates post-sensing and post-pacing AVD time intervals based on the one or more post-sensing AV, AQ*, AX, and AY time intervals and the one or more post-pacing AV, AQ*, AX, and AY

time intervals, respectively, as required for a DDD mode pacing. In one embodiment, memory circuit 1089 contains all the coefficients of the formulas used for the calculation of the AVD time intervals. In one embodiment, the coefficients are programmable. The user may enter new coefficients to replace the coefficients stored in memory circuit 1089. In one embodiment, the calculated AVD time intervals are also stored in memory circuit 1089. After at least one new AVD time interval is calculated, AVD calculator 1087 sends the new AVD time interval the therapy timing controller of implant controller 923 to control the timing of deliveries of ventricular pacing pulses. In one embodiment, AVD calculator 1087 is included in implant controller 923 and sends the new AVD time interval to the therapy timing controller portion of implant controller 923. In another embodiment, AVD calculator 1087 is included in programmer 840 and sends the new AVD time interval to the therapy timing controller of implant controller 923 via telemetry link 870.

In one embodiment, system 1080 includes software that is installed on programmer 840 when the AVD calculation, or at least a portion of the AVD calculation, is intended to be performed with that programmer. In one embodiment, the software constituting system 1080, or a portion thereof, is stored on one or more storage media allowing for installation when needed.

In one specific embodiment, the software installed in programmer 840 includes AVD calculator 1087. Programmer 840 also includes memory circuit 1089. Implant device 810 includes signal input 1081, event detector 1083, measurement module 1085. One or more of the post-sensing and post-pacing AV time intervals are telemetered from implanted device 810. AVD calculator 1087 calculates at least one the post-sensing and post-pacing AVD time intervals based on at least of the post-sensing and post-pacing AV time intervals, respectively. In general, this specific embodiment is suitable wherever implant device 810 is capable of detecting the events (any one or more of the A, V, Q*, X, and Y) and measuring the time intervals (any one or more of the AV, AQ*, AX, and AY).

In another specific embodiment, the software installed in programmer 840 includes signal input 1081, event detector 1083, measurement module 1085 and AVD calculator 1087. Programmer 840 also includes memory circuit 1089. Implanted device 810 senses the cardiac signals and telemeters at least one of the cardiac signals to programmer 840 tp provide for detection of at least one of the post-sensing and post pacing Q* events. Signal input 1081 receives the telemetered cardiac signal. Event detector 1083 detects the at least one of the post-sensing and post-pacing Q* events. Measurement module 1085 measures at least one of the post-sensing and post-pacing AQ* time intervals. AVD calculator 1087 calculates at least one of the post-sensing and post-pacing AVD time intervals based on the post-sensing and post-pacing AQ* time intervals, respectively. In general, this specific embodiment is suitable for calculating one or more AVD time intervals based on any of the AV, AQ*, AX, and AY time intervals wherever implant device 810 is capable of sensing and telemetering, without event detection, the one or more of the cardiac signals and/or the one or more of the mechanical signals.

Programming Adjustable AVD Time Intervals Using Look-up Table

In one embodiment, AVD time intervals are calculated for a list of AV, AQ*, AX, or AY time intervals or time interval ranges using the mathematical relationships discussed above. A look-up table is then produced to allow mapping of each AV, AQ*, AX, or AY time interval or time interval range to a calculated AVD time interval. This simplifies the programming by eliminating the need for calculations in clinical practice. In one specific example, the physician is provided with a look-up table allowing mapping of intrinsic AV intervals to AVD time intervals. The physician measures an AV time interval from one or more electrograms, finds the corresponding AVD time interval on the look-up table, and programming a pacemaker with that AVD time interval. This approach applies to determination of any post-sensing and post-pacing

AVD time intervals based on one of the post-sensing and post-pacing AV, AQ*, AX, and AY time intervals.

In a further embodiment, multiple lists of AVD time intervals are calculated, each based on a specific mathematical relationship developed for a group of patients classified by diagnosed clinical conditions including measured parameters indicative of the clinical conditions. In one specific embodiment, each list of the multiple lists of AVD time intervals is developed for patients having QRS duration within a particular range. The QRS duration is an indication of a degree of damage to a patient's cardiac conduction system. Studies have shown that QRS duration is one factor determining the patient's responsiveness to the resynchronization therapy and should be an factor determining the AVD time interval. The specific mathematical relationship is developed by determining the coefficients in the equations presented above empirically with patients having the QRS duration within that particular range. For example, equation $AVD = N1 \cdot AV - N2$ relates the AVD time interval to the AV time interval. Multiple sets of N1 and N2 are determined, each for patients showing a different QRS duration range. Thus, multiple specific equations are obtained relating AVD time intervals to AV time intervals for each particular QRS interval range.

FIG. 11 illustrates one embodiment of a look-up table 1100 for programming the adjustable AVD time interval. For the purpose of discussion with a specific example, look-up table 100 as shown in FIG. 11 relates (post-sensing) AVD time intervals to intrinsic (post-sensing) AV time intervals as one implementation of the methods for optimizing the AVD time interval as discussed above. It is to be understood that the methods and apparatuses discussed below in relation with a look-up table apply to implementations of all the methods for optimizing the AVD time interval discussed above, including using mathematical relationships each between the AVD time intervals and one of the AV, AQ*, AX, or AY time intervals.

Look-up table 1100 includes an input list 1110 of intrinsic AV time interval ranges (AV range 1, AV range 2, ..., and AV range N), another input list 1120 of QRS

duration ranges (QRS range 1 and QRS range 2), an output list 1130 of AVD time intervals each corresponding to one of the intrinsic AV time interval ranges and QRS range 1, and another output list 1140 of AVD time intervals each corresponding to one of the intrinsic AV time interval ranges and QRS range 2. For example, given that a patient's QRS duration falls within QRS range 1, and his/her intrinsic AV time interval falls within AV range 2, AVD time interval AVD1-2 is to be programmed for the patient.

In one embodiment, the AVD time intervals in look-up table 1100 include values at predetermined increments, such as in increments of 5 milliseconds or 10 milliseconds. In one embodiment, the AVD time intervals include minimum and maximum values for each of the QRS duration range. In one specific embodiment, QRS range 1 includes QRS durations within 120 to 150 milliseconds, and QRS range 2 includes QRS durations greater than 150 milliseconds. For QRS range 1, the corresponding AVD time intervals (AVD1-1 through AVD1-N) include values range from 50 to 250 milliseconds in 10 millisecond increments. For QRS range 2, the corresponding AVD time intervals (AVD2-1 through AVD2-N) include values range from 50 to 190 milliseconds in 10 millisecond increments.

In one embodiment, look-up table 1100 is presented in a visually readable form, such as on a card or in a user manual printed on paper. To program an AVD time interval, the physician measures an intrinsic AV time interval from an electrogram sensed with the electrodes used for pacing with the AVD time interval. The physician further measures the patient's intrinsic QRS duration. If a 12-lead ECG or other multi-vector ECG is recorded, the largest QRS duration among all recorded vectors is used as the measured QRS duration. The physician then identifies an AV range in list 1110 where the measured intrinsic AV interval falls into and a QRS range in list 1120 wherein the measured QRS duration falls into, and identifies an AVD time interval corresponding to the identified AV range and QRS range in list 1130 or 1140.

In a further embodiment, the AVD time interval identified from look-up table 1100 is adjusted before programming into implanted device 810. In one embodiment, pacing pulses are delivered to multiple ventricular sites with multiple AVD time intervals each used to time the pacing pulse delivery to one of the ventricular sites. In one specific embodiment, the method of determining the AVD time interval using look-up table 1100 is applied to determine one of the multiple AVD time intervals. The other AVD time intervals are each calculated by adding an offset value being an interventricular delay. In another embodiment, the AVD time interval is corrected by a sensed AV offset at the medical discretion of the physician. The sensed AV offset is to correct a delay in detecting a sensed intrinsic event from an electrogram.

In another embodiment, look-up table 1100 is presented on a screen of a computer or microprocessor based device. In one specific embodiment, look-up table 1100 is presented on display 941 of programmer 840 upon request. In another specific embodiment, look-up table 1100 is presented on a computer such as a laptop or palm computer used by the physician.

In another embodiment, look-up table 1100 is electronically stored in a microprocessor-readable form in a computer or microprocessor based device, such as programmer 840 or a laptop or palm computer used by the physician. In this embodiment, instead of identifying corresponding values in the visible form of look-up table 1100, the physician enters the measured intrinsic AV time interval and QRS duration through a user input device of the computer or microprocessor based device. The device maps the measured intrinsic AV time interval and QRS duration to the corresponding AVD time interval using the electronically stored look-up table 1100. In one embodiment, the corresponding AVD time interval is presented in an output device of the computer or microprocessor based device, such as a display screen.

FIG. 12 is a flow chart showing one method for producing an AVD look-up table such as look-up table 1100. A plurality of intrinsic AV value ranges (i.e., AV time interval ranges) are generated at 1200. In one embodiment, the AV value ranges are

large enough such that their corresponding AVD time intervals are practically distinctive. In one specific embodiment, the AVD time intervals are listed in predetermined increments such as increments of 5 or 10 milliseconds. The AV value ranges are generated in a way minimizing repeated values for the AVD time intervals in the look-up table. In another specific example, the AV value ranges are resulted by substantially evenly dividing an overall AV time interval range.

An AVD value (i.e., a value of the AVD time interval) for each AV value range is calculated based on a mathematical relationship between AVD and AV at 1210. The mathematical relationship provides for an approximately maximum positive rate of left ventricular pressure change during systole, LV+dp/dt, by delivering ventricular pacing pulses with the AVD. In one embodiment, the mathematical relationship is represented by an equation, $AVD = N1 \cdot AV - N2$, where N1 and N2 are coefficients empirically determined based data collected from patients suffering a particular class of conditions, and AV is an average AV value representing an AV value range. In one embodiment, a minimum value and a maximum value are given for the AVD values. The AVD value is set to the minimum value if the calculated AVD value is below the minimum value, and set to the maximum value if the calculated AVD value is above the maximum value. In one embodiment, step 1210 is repeated for at least a second mathematical relationship between AVD and AV. In one embodiment, the second mathematical relationship (as well as possible further mathematical relationships) is also represented by the equation, $AVD = N1 \cdot AV - N2$, where N1 and N2 are coefficients empirically determined based data collected from patients suffering another particular class of conditions. In one embodiment, the class of conditions are defined by a QRS duration. In one specific embodiment, the equation is $AVD = 0.7 \cdot AV - 55$ milliseconds (i.e., $N1 = 0.7$, $N2 = 55$ ms) for QRS durations above 150 milliseconds, and $AVD = 0.7 \cdot AV - 0$ milliseconds (i.e., $N1 = 0.7$, $N2 = 0$ ms) for QRS durations between 120 and 150 milliseconds. In one embodiment, the calculated AVD values are truncated or roundeded into AVD values in predetermined increments, such as in increments of 5 or

10 milliseconds. In one specific embodiment, the AVD time intervals listed on the look-up table are each a multiple of 5 or 10 milliseconds.

A look-up table allowing mapping the each AV value range to an calculated AVD value is produced at 1220. In one embodiment, this includes arranging the values and value ranges resulted from steps 1200 and 1210 according to the format illustrated by look-up table 1100 in FIG. 11, and presenting the look-up table on a card, a user's manual, or a display screen. In another embodiment, this includes storing the values and value ranges resulted from steps 1200 and 1210 and the mapping relationships between these values on a computer-readable medium, and providing a device that receives parameters measured by the physician and returns the corresponding AVD time interval mapped using the look-up table.

Conclusion

The present pacing system may be employed in a variety of pacing devices, including implantable pacing devices. The present system may be used for pacing one or more ventricles. A variety of pacing electrode configurations may be employed without departing from the present invention including multiple pacing sites at a ventricle(s), provided that the required electrical or mechanical events are monitored. Changes in the coefficients and order of methods provided herein may be practiced accordingly without departing from the scope of the present invention.